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An EMG technique for measuring spinal loading during asymmetric lifting

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Abstract

Objective. To compare two methods of calibrating the erector spinae electromyographic signal against moment generation in order to predict extensor moments during asymmetric lifting tasks, and to compare the predicted moments with those obtained using a linked-segment model.

Methods. Eight men lifted loads of 6.7 and 15.7 kg at two speeds, in varying amounts of trunk rotation. For each lift, the following were recorded at 60 Hz; the rectified and averaged surface electromyographic signal, bilaterally at T10 and L3, lumbar curvature using the 3-Space Isotrak, movement of body segments using a 4-camera Vicon system, and ground reaction forces using a Kistler force-plate. Electromyographic (EMG) and Isotrak data were used to calculate lumbosacral extensor moments using the electromyographic model, whereas movement analysis data and ground reaction forces were used to estimate net moments using the linked-segment model. For the electromyographic technique, predictions of extensor moment were based on two different sets of EMG-extensor moment calibrations: one performed in pure sagittal flexion and the other in flexion combined with 45° of trunk rotation.

Results. Extensor moments predicted by the electromyographic technique increased significantly with load and speed of lifting but were not influenced by the method of calibration. These moments were 7–40% greater than the net moments obtained with the linked-segment model, the difference increasing with load and speed.

Conclusions. The calibration method does not influence extensor moments predicted by the electromyographic technique in asymmetric lifting, suggesting that simple, sagittal-plane calibrations are adequate for this purpose. Differences in predicted moments between the electromyographic technique and linked-segment model may be partly due to different anthropometric assumptions and different amounts of smoothing and filtering in the two models, and partly due to antagonistic muscle forces, the effects of which cannot be measured by linked-segment models.

Relevance

Asymmetric lifting is a significant risk factor for occupationally-related low back pain. Improved techniques for measuring spinal loading during such complex lifting tasks may help to identify work practices which place the spine at risk of injury.

Keywords: Lumbar spine; Compressive loading; Electromyography; Linked-segment model; Asymmetric lifting

1. Introduction

High spinal loading during manual handling is mostly attributable to forces generated by the muscles of the back and abdomen, so it is an attractive idea to attempt to quantify these forces directly from the electromyographic (EMG) activity of the trunk muscles. This direct approach has two advantages over more traditional techniques such as linked-segment modelling: firstly, EMG systems are portable and suitable for use in the workplace; secondly, they have the potential to measure antagonistic muscle forces which cannot be detected by the linked-segment models. However, direct EMG measurements are inherently noisy, and the EMG-force relationship is influenced by several variable factors such as muscle length [1–4] and contraction velocity [1,5,6].

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In previous work, we have attempted to overcome the difficulties associated with making direct EMG estimations of spinal loading [1]. The movement artefact in EMG signals is minimised by reducing the skin's electrical resistance below 10 kΩ, by high pass filtering, and by averaging and smoothing signals from several muscle sites. Static EMG-extensor moment calibrations are performed in a range of different postures to account for the influence of changes in muscle length [1,2], and global correction factors obtained from isokinetic calibrations are applied to EMG signals recorded from shortening muscles in order to account for the influence of contraction velocity and electromechanical delay [1,7]. A potential difficulty that remains with surface EMG is that the signal detected from superficial muscles will dominate the surface EMG signal. This effect can be partly overcome by using large surface electrodes that detect signal from a large volume of muscle. However, deep muscle signal will still be under-represented. In the technique used in this paper, this may not be a problem because the predictions of extensor moment are based on the relationship between the *detected* signal and the extensor moment. Therefore, provided the relative contribution from deep muscles to the detected signal is similar in the dynamic lifts to that in the calibrations then estimates of extensor moment will be little affected. Another problem with the EMG technique is that trunk muscle EMG measurements are fundamentally incapable of detecting forces generated by muscles of the legs and pelvis which accelerate the spine axially, i.e. in the direction of its long axis. Such forces will not produce any bending about the L5–S1 joint of the spine and so will not be evident in the measured extensor moment, but they will add to spinal compression. However, these "hidden" inertial forces have been shown to be comparatively small during moderate lifting tasks, so that even during rapid lifting movements with bent knees, they rarely exceed 2–4% of the total spinal compressive force [8].

One further problem associated with the direct EMG estimation of spinal loading is tackled in the present paper. Occupational lifting often requires weights to be lifted while the spine is positioned in some awkward rotated posture, rather than in the sagittally symmetric postures considered previously in our laboratory. In such postures, asymmetries in muscle activity [9–11] and marked reductions in strength [12–14] may influence the EMG-extensor moment relationship and hence the predictions of the EMG model. In the present study, therefore, we extend our EMG techniques to measure spinal loading during asymmetric lifting. The main purpose of the investigation is to compare the effects of asymmetries in trunk muscle activation patterns on EMG-based predictions of extensor moment and spinal compression. However, we also had the opportunity to compare the EMG model predictions with those obtained simultaneously using a fully dynamic three-dimensional linked-segment model [15]. The three-dimensional data from the linked-segment model have been published previously in a separate study which examined whether pelvic twist prevents asymmetric loading of the spine [16]. In this study, we used the sagittal-plane data from the linked-segment model to make comparisons with the EMG-based measurements of spinal loading.

2. Methods

2.1. Subjects

Eight healthy men with no history of low back pain gave their informed consent to participate in the study. Their mean age was 26 yr (range 21–36 yr) and their average body mass was 72 kg (range 63–81 kg).

2.2. Lifting tasks

Subjects lifted a box with handles on either side, in varying amounts of right-sided rotation (0°, 10°, 30°, 60° and 90°) from the sagittal midline, as defined by the position of the feet. At 0°, the box was placed symmetrically in front of the feet, and at 90° it was in line with the shoulders in the frontal plane (Fig. 1(a)). In each lift, the box was lifted from a shelf 10 mm above the ground, finishing in an upright symmetrical standing position (Fig. 1(b)). Lifts were performed using two different loads (6.7 and 15.7 kg) and two different speeds, in time with a metronome: the "slow" lift was completed in approximately 1.5 s and the "fast" lift in 1.0 s. All lifts were performed twice, and the order was randomised to avoid any systematic effects of training or fatigue. Subjects were asked to perform the lifts freestyle, and to adopt a comfortable starting position with their hands gripping the box handles. Before the first lift, the distance between the greater trochanter of their left femur and the ground was measured. This enabled the starting position for each subject to be standardised for all subsequent lifts to ensure that the same amount of knee flexion was used each time. An electrical switch attached to the base of the box was activated when the box left the ground, and this signal was used as a synchronisation signal that defined time "zero" for all measurements.

2.3. Estimation of lumbosacral extensor moments using the EMG technique

Full details of this technique have been published previously [1]. In this study, four pairs of surface electrodes (Biolect, UK) were used to record the EMG signal bilaterally at the levels of T10 and L3, and a

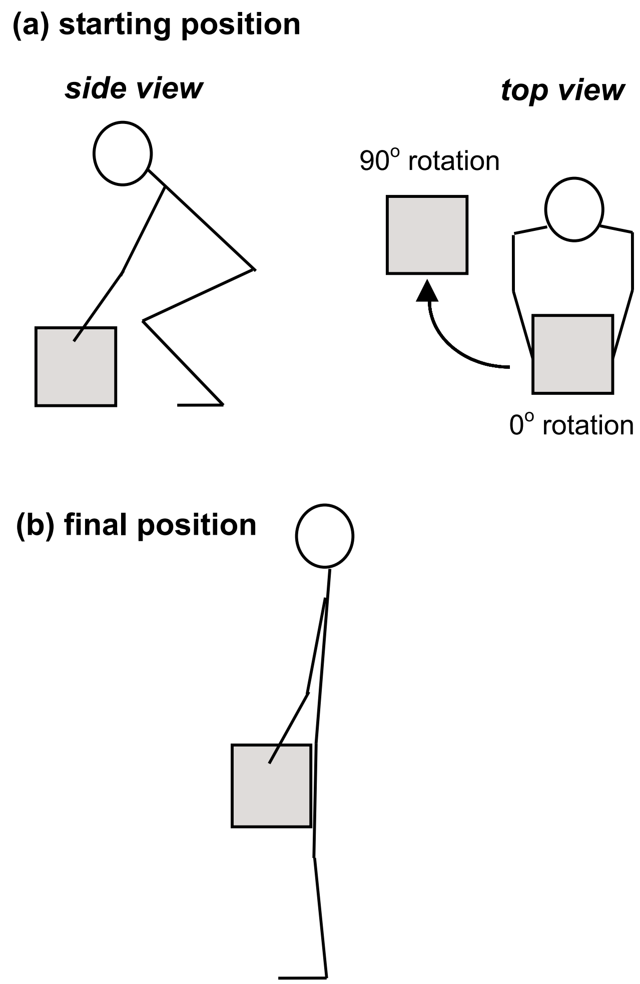

Fig. 1. Subjects began each lift with the knees flexed and the load placed in varying amounts of right-sided rotation (0°, 10°, 30°, 60° or 90°) from the sagittal midline (a). Subjects completed each lift to finish in an upright symmetrical standing position (b).

reference electrode was placed on the sternum. Careful skin preparation, which involved abrading the skin and cleaning it with alcohol, ensured that the impedance between the reference electrode and each of the recording electrodes was less than 10 kΩ, and that the signal-to-noise ratio was high. The EMG signal from the erector spinae muscles was recorded throughout each lift, and pre-amplified by separate isolation pre-amplifiers in order to minimise mains intereference on the signal. The signal was then conveyed to the main amplifier where it was further amplified in the range 8–300 Hz, full-wave rectified and averaged with a time constant of 0.05 s, and A-D converted at 60 Hz for subsequent storage on a PC. Simultaneous recordings were also made of lumbar curvature, using the 3-Space Isotrak (Polhemus, Colchester, VT). This is an electromagnetic tracking device which consists of a source of electromagnetic waves, that was attached to the skin overlying the sacrum, and a small sensor of these waves, that was attached to the skin overlying the L1 spinous process. The angle between the sensor and source indicates the lumbar curvature [17]. During each lift, changes in lumbar curvature were recorded to allow the dynamic EMG signal to be corrected for changes in muscle length (indicated by lumbar curvature), and contraction speed (indicated by rate of change of lumbar curvature) [1].

Calibration of the EMG signal against extensor moment was carried out during two sets of isometric contractions. These were performed with the subject positioned in a stabilisation frame in which the pelvis could be immobilised. Subjects flexed forward by a predetermined amount, keeping their arms straight, and grasped a handlebar attached by a variable length chain to a floor-mounted load cell. They were then asked to pull upwards on the handlebar with increasing force in order to reach a maximum value after 3 s. During each pull, the lumbar curvature was measured using the 3-Space Isotrak. In addition, the EMG activity of the erector spinae, averaged over the four muscle sites, and the load exerted on the load cell, were recorded at 60 Hz. The total extensor moment exerted throughout the pull was calculated by adding the active moment, due to the load exerted on the load cell, to the moment exerted by the upper body. The latter was calculated by assuming that upper body mass above L5–S1 was equivalent to 60% of total body mass, and that its centre of mass acted at a point that lay 60% of the horizontal distance between the approximate centre of the L5–S1 disc and the centre of the handlebar [1]. Linear regression analysis was then used to determine the relationship between the averaged EMG and the total extensor moment relative to the centre of the L5–S1 disc [1].

In the first set of calibrations, subjects performed between six and eight isometric contractions, each in a different amount of sagittal flexion but without any rotation. Lumbar curvature was randomly varied, at approximately 10% intervals, in order to obtain values between 40% and 100% of the range between erect standing and full lumbar flexion. In the second set of calibrations (performed by seven of the eight subjects), the isometric contractions were repeated with the trunk and shoulders rotated 45° to the right from the midsagittal plane as defined by the position of the pelvis. This amount of rotation (relative to the pelvis) approximates to the limit of thoracolumbar rotation indicated by direct in vivo measurements obtained by inserting pins into the spinous processes of healthy volunteers [18]. The purpose of using calibration data obtained at 0% and 100% of the range of trunk rotation was to increase the likelihood of detecting any effect of trunk rotation on the predicted moments. For each set of calibrations (non-rotated or rotated) performed by each subject, regression analysis was then used to express the gradient and intercept of the EMG-extensor moment relationship as a function of muscle length (lumbar curvature), as indicated by the Isotrak measurements [1].

In order to convert the EMG signals recorded during the dynamic lifting activities into extensor moments, the recorded signal was first corrected for the effects of electromechanical delay and contraction speed (rate of change of lumbar curvature) using average correction factors obtained in a previous study [1]. The corrected EMG (E_0) was then compared with the isometric calibrations using the following equation:

$$\text{Extensor moment} = E_0 G + I, \tag{1}$$

where G and I are the gradient and intercept of the EMG-extensor moment relationship, predicted from the isometric calibrations as variable functions of lumbar curvature. The intercept, I, indicates the moment that is generated when there is no muscle activity, and is attributable to tensile forces in non-contractile tissue in the muscle and in other structures such as fascia, disc and ligaments. Calculated moments therefore represent the combined effects of active muscle contraction and passive tension in non-contractile tissues.

Extensor moments were calculated using values of G and I obtained from non-rotated and rotated calibrations, and in each case the peak extensor moment during the lift was determined. In the case of the rotated calibrations, the "extensor moment" was calculated in the rotated sagittal plane of the shoulders, and not in the sagittal plane as determined from the position of the feet and pelvis. It should also be noted that although abdominal muscle activity was not recorded directly in the present study, the increase in extensor moment that would occur as a result of such flexor muscle activity would be detected by the EMG technique employed.

2.4. Estimation of net lumbosacral moments using the linked-segment model

Full details of this technique have been reported previously [15]. During each lift, the vertical ground reaction force was recorded at 60 Hz using two Kistler force plates, one under each foot. The movement of different body segments was determined using a 4-camera Vicon system which recorded the positions of reflective markers attached to the feet, lower legs, upper legs and pelvis via thermoplastic cuffs which could be moulded to the contours of each individual [15]. Segment inertial parameters (i.e. mass, centre of mass location and inertia tensor) of the feet, lower legs, upper legs and pelvis were calculated using regression equations developed by McConville et al. [19]. This anthropometric data together with the ground reaction forces recorded by the Kistler force-plates, and the motion analysis data recorded by the Vicon system were input into a fully-dynamic 3-dimensional linked-segment model in order to estimate the net sagittal moment acting at the lumbosacral joint throughout each lift. The word "net" indicates that the measured moment is the extensor moment minus any flexor moment generated by antagonistic contractions of the abdominal muscles. The "sagittal plane" referred to in the linked-segment model is that defined by the (rotated) position of the pelvis rather than that defined by the (rotated) position of the shoulders as in the EMG model. The linked-segment model was also used to determine net moments at L5-S1 in the frontal and transverse planes. In this study, moments were calculated using a "bottom-up" algorithm, and the peak moments during each lift were determined. This model has been validated previously by comparing the lumbosacral moments predicted by "bottom-up" and "top-down" analyses, and these showed good agreement [15].

2.5. Statistical analysis

A four factor repeated measures analysis of variance was used to compare the effects of load, lifting speed, amount of rotation, and repetition on the "sagittal" moments predicted by each separate model: (a) EMG technique (non-rotated calibrations); (b) EMG technique (rotated calibrations); (c) linked-segment model. Peak moments obtained with the EMG model using non-rotated calibrations were also compared independently with those obtained using the rotated calibrations, and with those obtained using the linked-segment model, using a four factor repeated measures analysis of variance where the model replaced the repetition as the fourth independent variable. Significance was accepted at the 5% level. Mean ($\pm$SD) values are shown throughout.

3. Results

3.1. EMG model (non-rotated calibrations)

Using the non-rotated calibration data, predicted peak lumbosacral extensor moments ranged from 181 ($\pm$53) Nm for the slow lift of 6.7 kg, performed in 90° of trunk rotation, up to 330 ($\pm$105) Nm for the fast lift of 15.7 kg, performed in 0° of rotation (Table 1). Using appropriate internal lever arms for the erector spinae muscles [1,20], these extensor moments corresponded to compressive forces of 2.5–4.8 kN. Peak extensor moments increased significantly with both speed ($P = 0.013$) and load ($P < 0.001$) but were not affected by the amount of trunk rotation. However, there was a significant interaction between the effects of speed and rotation ($P = 0.025$). Peak extensor moments did not

Table 1
Peak lumbosacral *extensor* moments predicted by the EMG technique using sagittal plane calibration data (EMG_a), and peak lumbosacral *net* moments predicted by the linked-segment model (LSM_c), for each of the lifts studied[a]

Load/speed	Rotation of the load from the sagittal midline									
	0°		10°		30°		60°		90°	
	EMG_a	LSM_c	EMG_a	LSM_c	EMG_a	LSM_c	EMG_a	LSM_c	EMG_a	LSM_c
6.7 kg slow	225 (64)	183 (21)	233 (66)	187 (23)	207 (59)	181 (22)	190 (48)	175 (23)	181 (53)	164 (24)
6.7 kg fast	260 (77)	211 (27)	248 (88)	214 (29)	267 (89)	209 (31)	237 (66)	195 (28)	221 (81)	185 (32)
15.7 kg slow	283 (77)	233 (30)	292 (72)	235 (33)	256 (72)	230 (30)	255 (57)	218 (32)	252 (85)	201 (38)
15.7 kg fast	330 (105)	256 (38)	326 (97)	258 (36)	307 (83)	250 (35)	296 (98)	233 (32)	301 (104)	217 (33)

[a] Values shown are the mean (±SD) based on two repetitions of each lift for eight healthy male subjects.

differ significantly between the first and second repetition of each lift suggesting that there were no systematic alterations in lifting technique at the second attempt.

3.2. EMG Model (rotated calibrations)

Predicted peak extensor moments based on the rotated calibrations ($n = 7$) ranged from 194 (±32) Nm up to 333 (±117) Nm. Predicted moments showed the same trends as those based on the non-rotated calibrations, with load ($P < 0.001$) and speed ($P = 0.04$) being the only factors to have a significant effect upon them.

3.3. Linked-segment model

The linked-segment model predicted peak lumbosacral moments in the sagittal plane (defined by the pelvis) that ranged from 164 (±24) Nm for the slow lift of 6.7 kg, performed in 90° of rotation, up to 258 (±36) Nm for the fast lift of 15.7 kg, performed in 10°of rotation (Table 1). Peak moments increased significantly with speed ($P < 0.001$) and load ($P < 0.001$) but decreased with increasing angle of trunk rotation ($P = 0.013$). Within subject contrasts indicated a significant reduction in peak moment at each level of torsion ($P < 0.001$). There was a good agreement in peak moments between the first and second repetitions of each lift such that no significant differences were found between them.

3.4. Comparison of EMG model predictions using non-rotated and rotated calibrations

There were no significant differences in peak moments when values obtained using the rotated calibrations were compared with those obtained using the non-rotated calibrations (Fig. 2), regardless of the load, speed or

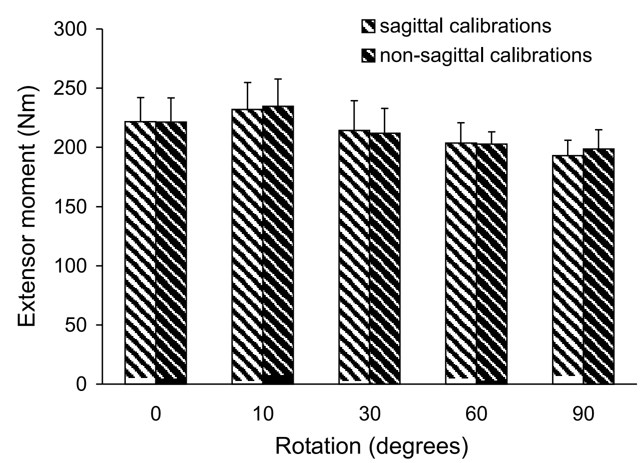

Fig. 2. Peak lumbosacral extensor moments during slow lifts of 6.7 kg performed in varying amounts of rotation from the sagittal midline. Extensor moments were calculated using the EMG model based on sagittal plane and non-sagittal plane calibration data. Mean values are shown for seven healthy men. Bars indicate the standard error of the mean.

amount of rotation. The greatest mean difference of $4.0 \pm 11\%$ was observed for the fast lift of 6.7 kg in 90° of rotation. For all other lifts, the mean difference was usually less than 2%.

3.5. Comparison of EMG model and linked-segment model predictions

Extensor moments predicted by the EMG model, based on the non-rated calibrations, were significantly greater than the net sagittal moments (in the plane of the pelvis) predicted by the linked-segment model ($P < 0.05$). On average, the EMG predictions were 21% higher than those of the linked-segment model, and differences between the two techniques increased with

Table 2
Differences in peak lumbosacral moments between the EMG technique (EMG_a) and the linked-segment model (LSM_c) for each of the lifts studied[a]

Load/speed	Rotation of the load from the sagittal midline										Average	
	0°		10°		30°		60°		90°			
	Nm	(%)	Nm	(%)	Nm	(%)	Nm	(%)	Nm	(%)	A	B
6.7 kg slow	42	21.7	46	23.9	26	13.4	15	7.4	17	9.2	29	15.1
6.7 kg fast	49	22.8	34	15.8	57	27.1	42	20.5	36	16.8	44	20.6
15.7 kg slow	50	21.5	57	23.7	27	10.6	38	16.8	65	30.1	47	20.5
15.7 kg fast	74	29.9	68	26.1	57	22.4	63	25.2	91	39.8	71	28.7
Average	53	23.7	49	21.5	43	18.6	43	19.0	54	24.8	48	21.5

[a] Values shown are the mean based on two repetitions of each lift for eight healthy male subjects. A = difference in Nm, B = % difference between the EMG and LSM moments.

speed and load (Table 2). Matched pair t-tests indicated that the magnitude of the differences between the models was significantly greater at the higher of the two loads ($P < 0.005$) and the faster of the two speeds ($P < 0.002$). Average differences in predicted moments ranged from 15 Nm (7% of the linked-segment model net moment) in slow lifts of 6.7 kg up to 91 Nm (40%) in fast lifts of 15.7 kg (Table 2 and Fig. 3).

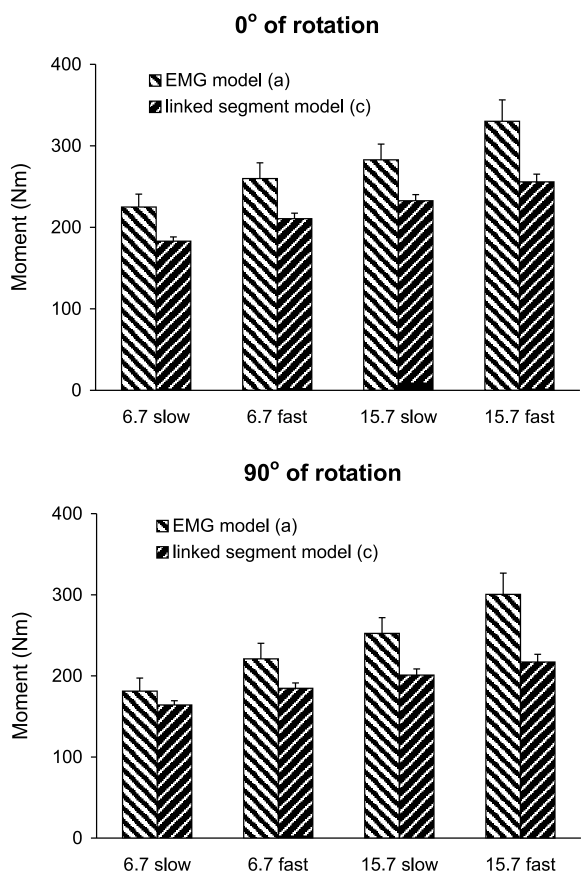

Fig. 3. Peak lumbosacral moments during lifts performed in 0° (upper) and 90° (lower) of rotation from the sagittal midline. Mean values are shown for each of two repetitions in eight healthy men. Bars indicate the standard error of the mean.

4. Discussion

The results of this study show that EMG-based estimates of peak extensor moments generated during lifting activities are little affected by trunk asymmetries during the static EMG-moment calibrations. Almost the same predictions of peak extensor moment were obtained when the EMG calibrations were performed without trunk rotation, or with it, and this was equally applicable to dynamic lifts performed in a range of rotated postures. This result may appear to be at variance with previous findings that asymmetric loading leads to large increases in erector spinae muscle activity on the side contra-lateral to the direction of twisting or lateral bending [9–11]. However, there is no fundamental problem here: the present results merely indicate that increased muscle activity on the lengthened contra-lateral side of the back is accompanied by equivalent reductions on the shortened ipsilateral side, so that the bilateral EMG activity averaged over the four recording sites is similar to that generated during sagittally symmetric lifts. The practical implication of this finding is that EMG techniques can be used to quantify extensor moments during asymmetrical lifts, even when the EMG-moment calibrations are performed only in non-rotated positions.

One of the fundamental problems with estimates of spinal loading is that there is no absolute "gold standard" by which to judge the measurements obtained. However, several aspects of the present results give a certain amount of confidence in the validity of both the EMG and linked-segment model techniques. Firstly, the two techniques give broadly similar estimates of peak spinal loading (Table 1) even though they are based upon entirely different assumptions and approximations. While this does not prove that either technique is accurate, it does give increased confidence that both are without large systematic errors. Secondly, the ability of both techniques to demonstrate consistent and significant increases in spinal loading resulting from changes in the weight lifted and lifting speed, confirm the previous results obtained in symmetric [1,7,21,22] and asymmetric lifting [23,24].

Consistent differences in predicted moments between the EMG technique and linked-segment model, which averaged 21% across the range of lifts studied, can be explained in terms of different anthropometric assumptions, differing degrees of smoothing and filtering of raw data, and differences in the models' abilities to assess the effects of antagonistic muscle activity. The relative importance of these factors is discussed in detail elsewhere [25]. However, our calculations suggest that anthropometric assumptions relating to the relative mass of the upper body and legs account for about 6–8% of the difference between the models, and antagonistic muscle activity accounts for a further 5–10%. The fact that the *discrepancies* in predicted moments between the techniques changed in a systematic fashion with lifting speed and load (Fig. 3) may be attributable to an increase in antagonistic stabilising forces under these circumstances. Such increases in antagonistic muscle activity with increased severity of loading have been demonstrated previously in studies that directly measured the activity of abdominal muscles [6,26,27].

Any remaining differences between the models may be due to smoothing and filtering effects [25] or to the different axes about which moments were calculated in the two models.

The extent of this latter effect cannot be determined directly because the position of the trunk and shoulders was not recorded in the lifts. However, if the deviation in the axis systems used by the two models increased with increasing rotation in the lifts then this would be reflected in the linked-segment model predictions by an increase in the lumbosacral twisting moment and an increase in the total moment relative to the sagittal moment (in the plane of the pelvis). Differences between the total moment and the sagittal moment estimated by the linked-segment model ranged from 2 Nm (1%) with 0° of box rotation up to 29 Nm (15%) with 90° of box rotation [16], and in all cases the contribution from rotation to the total moment was less than that from lateral flexion. However, even assuming that these contributions were equal, the maximum difference in the total moment due to rotation of the trunk relative to the pelvis would be 15 Nm or 7.5% of the sagittal moment predicted by the linked-segment model. Based on these calculations, it is likely that differences in the sagittal axes used by the EMG and linked-segment models would result in small differences between their predicted moments in the range of 1–7.5% of the linked-segment model estimates. However, despite these calculations, there was in fact no significant effect of rotation on the *differences* between the two models. The linked-segment model did predict a significant reduction in the sagittal moment with increasing trunk rotation, and although a similar trend was observed for the EMG model estimates, this was less marked and failed to reach significance, possibly because of the differences in the measurement axes discussed above. However, it is worth noting that while the sagittal moment estimated by the linked-segment model decreased with increasing asymmetry the total net moment at the lumbosacral joint remained almost constant as trunk rotation increased [16], indicating that there was no overall reduction in spinal loading. These findings suggest that there is increased activity of abdominal muscles with increasing trunk rotation which contributes to increases in the lateral flexing and twisting torques, as indicated by previous studies [6,27].

5. Conclusions

The EMG technique used in this study is suitable for assessing dynamic spinal loading during asymmetric manual handling tasks. The use of more complex asymmetric EMG calibrations had no significant effect on the predicted extensor moments suggesting that simple sagittal calibrations are adequate even for complex lifting tasks. The EMG and linked-segment models give broadly similar estimates of peak spinal loading even though they are based upon entirely different assumptions and approximations, and both models are sufficiently sensitive that they can detect consistent and significant increases in spinal loading resulting from changes in lifting conditions. Differences between the models may be explained in terms of different anthropometric assumptions, differing degrees of smoothing and filtering of raw data, differences in the sagittal axes and differing abilities to measure the effects of antagonistic muscle activity.

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